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CONFERENCE PROCEEDINGS

How to Interpret Ventilator Waveforms Using the Taxonomy for Modes of Mechanical Ventilation

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Abstract

Mechanical ventilation is a complex yet essential aspect of critical care. This article provides a brief review of historical developments related to descriptions of ventilators, the derivation of the foundational equation of motion for the respiratory system and its derivatives, and a short overview of the taxonomy for modes of ventilation. These topics are then applied in a step-by-step procedure for interpreting patient–ventilator interactions through waveform analysis.

Keywords: mechanical ventilation, equation of motion, ventilator modes, taxonomy, patient–ventilator interaction, waveform interpretation, respiratory mechanics, patient–ventilator asynchrony

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Introduction

Ventilator waveform interpretation is analogous to electrocardiogram (ECG) interpretation, a fundamental skill for deciphering the heart's electrical activity and its physiological implications. Interpretation of ECG waveforms requires first an understanding of the physiology of the heart that is reflected in the data. Next, the appearance of the expected waveform (ie, normal sinus rhythm) is memorized and used as a comparator to deviations caused by physiological abnormalities (atrial flutter, atrial ventricular (AV) blocks, etc). A stepwise procedure is often used to generate a formal ECG interpretation (rate, rhythm, P-wave analysis, PR interval, QRS complex, etc).

This rigorous approach to ECG interpretation stands in stark contrast to our current practices with mechanical ventilation. Although we routinely observe ventilator displays and document settings, there's a notable absence of systematic data collection regarding waveform interpretation. Although some clinicians consistently analyze ventilator waveforms to assess the status of patient–ventilator interactions, the practice remains an exception rather than the norm. This disparity highlights a significant opportunity to enhance our understanding and management of mechanically ventilated patients.

The purpose of this article is to establish a theoretical framework for ventilator waveform interpretation as a basis for understanding patient–ventilator discordance. Waveform interpretation involves three basic steps: (1) identification of the mode of ventilation, (2) recognition of any distortion from expected waveform appearance because of abnormal elastic or resistive loads or effects of patient breathing efforts, and (3) classification of the patient–ventilator interaction as normal or abnormal according to the phase of the breath cycle (ie, trigger, inspiration, cycle, expiration). This article will deal with the first 2 steps. The third step has been treated elsewhere.[1]

The first step, recognition of the mode of ventilation (ie, a pre-set pattern of patient–ventilator interaction) from the appearance of the waveform only, is complicated by the fact that there are hundreds of brand names for modes just as there are thousands of brand names for drugs. Therefore, we must begin by understanding the taxonomy for modes of ventilation for the same reasons we must use generic names for drugs. The second step, recognition of elastic and resistive loads, requires more than simple pattern recognition: There must be some understanding of the basic physics and mathematical models used to describe patient–ventilator interactions. In my experience, the main reason that clinicians cannot deduce patient–ventilator interactions simply by looking

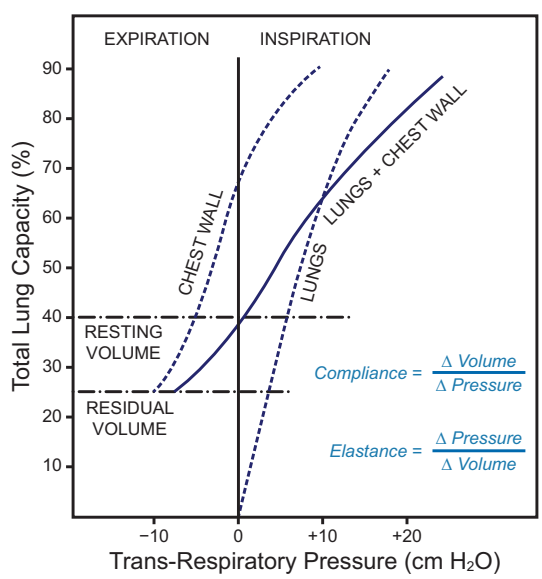


Fig. 1. Classic definitions of compliance and elastance.

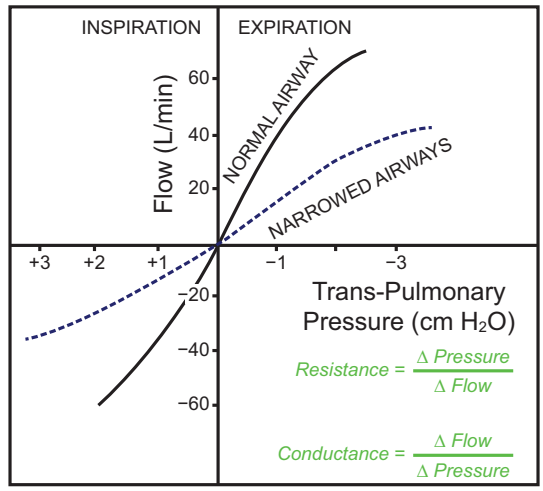


Fig. 2. Classic definitions of resistance and conductance.

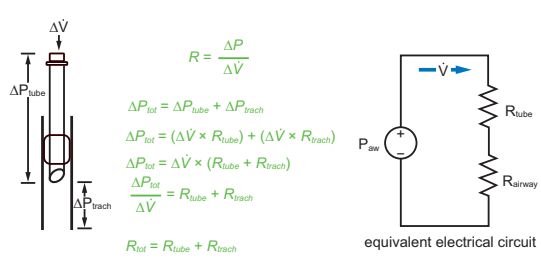


Fig. 3. Resistances connected in series. R = resistance, P = pressure, $\dot{V}$ = flow.

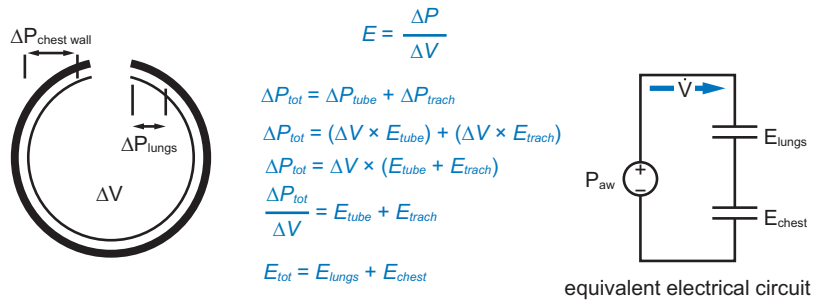


Fig. 4. Elastances connected in series. E = elastance, V = volume.

at waveforms is that they lack the requisite mathematical background. Although mastery would require competence in calculus, expert clinical practice requires little more than high school mathematics. But it is more than that. Practicing the application of the theory under the guidance of a mentor is essential. If you want to learn to fly, you begin in flight school. But nothing takes the place of flying a real airplane (or simulator).

To address these topics, we will begin with some historical context.

Early Respiratory Therapy Equipment Books

Early textbooks played a crucial role in disseminating knowledge about ventilators. In 1995, a significant advancement occurred with the publication of a new respiratory care equipment book.[2] This particular book was groundbreaking because it was the first to use the *equation of motion for the respiratory system* (described in the next section below) as the theoretical basis for classifying modes of ventilation. Furthermore, it provided equations that explain the waveforms displayed on mechanical ventilators, serving as a foundation for software simulators.

Remarkably, classic textbooks used in the early studies of respiratory care did not mention the equation of motion.[3–6] However, these texts did describe fundamental mechanical properties such as compliance (and its reciprocal, elastance), as well as flow resistance (and its reciprocal, conductance). These are illustrated in Figures 1 and 2.

Detailed models of the respiratory system are composed of more than a single resistance and compliance. Accordingly, these textbooks showed how to calculate the total effective resistance and compliance for individual mechanical components connected in series and parallel configurations (Figs. 3–8). *Note: Components connected in series share the same flow but have different pressure drops across them. Components connected in parallel have the same pressure drop across them but have different flows.*

The take-home message is that resistors in series add like compliances in parallel. This means that the total value is more than each individual value. But resistors in parallel add like compliances in series. In this case, the total value is less than each individual value.

Conversely, other classic textbooks from the 1980s, 1990s, and later *did* describe the equation of motion.[7–10] The most recent edition (13th) of "Egans Fundamentals of Respiratory Care" provides a detailed description of the equation and its support for mode taxonomy.[11]

The Equation of Motion

The equation of motion (EoM) is an essential mathematical model for understanding mechanical ventilation, to

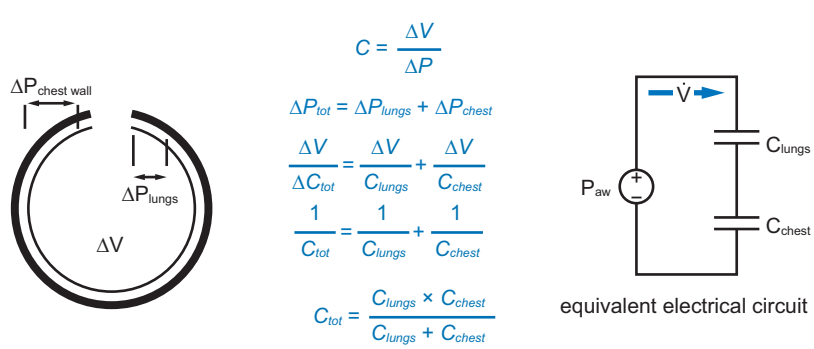


Fig. 5. Compliances connected in series. C = compliance.

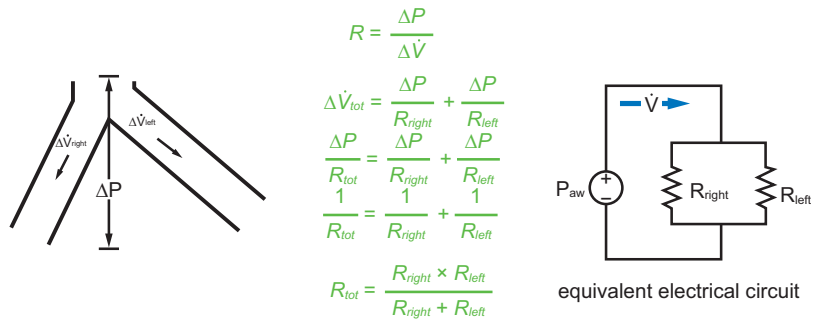


Fig. 6. Resistances connected in parallel.

the extent that one cannot become an expert without at least an intuitive grasp of it. Its development forms the central theme for understanding ventilator design, modes, and mechanical patient–ventilator interactions. In the English language, the first use of the EoM was likely by Otis, Fenn, and Rahn in 1950, who derived it to support their research into unassisted normal breathing.[12] In 1961, Jere Mead later popularized the equation for understanding mechanical ventilation, demonstrating the utility of electrical analogs and circuit analysis equations for comprehending ventilator performance.[13] Marini and Crooke further expanded its use through a series of articles, describing a general EoM applicable to any mode of ventilation, initially for passive ventilation and later expanded to include active ventilation, where patient inspiratory efforts interact with the ventilator's output.[14–18]

The EoM simplifies the complex mechanical behavior of the respiratory system by using a single-compartment model, which allows description of patient–ventilator interaction in terms of only 3 variables: pressure, volume, and flow (Fig. 9).

The term pressure itself can be confusing.[19,20] It may refer to absolute pressure (ie, relative to a vacuum), gauge pressure (ie, relative to atmospheric pressure), or pressure relative to a clinical reference such as PEEP (eg, inspiratory pressure above PEEP). It can also refer to a difference between 2 points in space (eg, trans-respiratory, trans-pulmonary) or 2 points in time (eg, driving pressure, occlusion pressure), or even between 2 points in space and time. These differences are shown in Table 1.

Common point pressures and pressure differences for the respiratory system are shown in Figure 10.[19] Indeed, it can be said that the "structures" that exist in the respiratory system are defined as whatever exists in the space between the points of pressure in these pressure differences. Understanding these distinctions is crucial for bedside calculations and comprehending ventilator operation.

The EoM can be derived using electrical circuit analysis by modeling the patient–ventilator system with electronic symbols (Fig. 11) using Kirchhoff's Voltage Law. Kirchhoff's Voltage Law says that if you add up all the voltage increases and decreases around any closed loop in an electrical circuit, the total will always be zero. In simple terms, the energy supplied to a circuit loop exactly equals the energy used by the components in that loop.

Alternatively, we can derive the EoM using Newton's third law of motion, which states that for every action

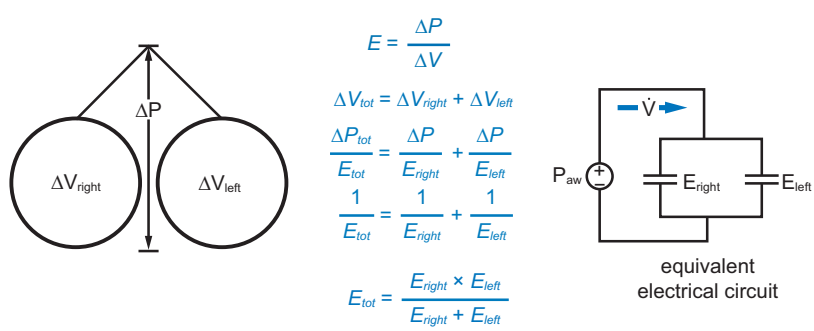


Fig. 7. Elastances connected in parallel.

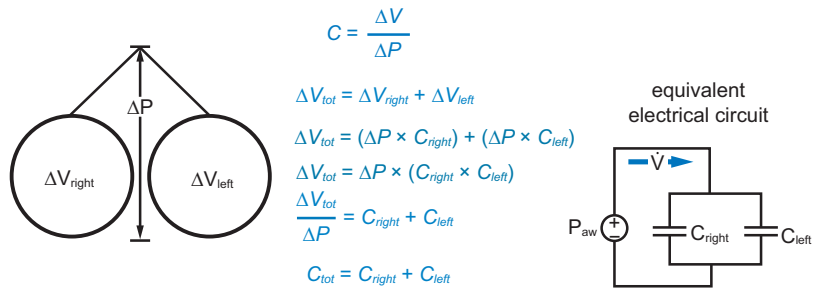


Fig. 8. Compliances connected in parallel.

there is an equal and opposite reaction. In terms of pressures, we can say:

$$P_{action}(t) = P_{reaction}(t) \tag{1}$$

Expanding on this idea, the gauge pressure at the airway opening that causes inspiratory flow is impeded by the elastic and resistive pressures of the respiratory system, which are variable functions of time. We must also include any constant pressures such as PEEP, either set on the ventilator or intrinsic because of gas trapping:

$$P_{aw}(t) = P_E(t) + P_R(t) + PEEP_{set} + PEEP_{auto} \tag{2}$$

where

P_{aw} = pressure at the airway opening
P_E = pressure due to elastic recoil
P_R = pressure due to flow resistance
set PEEP = PEEP set on the ventilator
autoPEEP = autoPEEP (also known as intrinsic PEEP)
t = time

Equation 2 is then expanded using the definitions for elastance and resistance:

$$P_{aw}(t) = E \cdot V(t) + R \cdot \frac{dv}{dt} + PEEP_{tot} \tag{3}$$

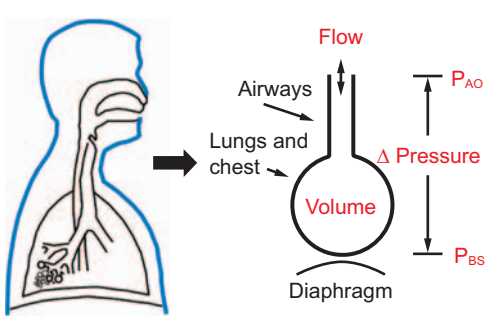


Fig. 9. The single-compartment model of the lungs as the basis of the most common form of the EoM. P_{AO}, pressure at the airway opening; P_{BS} = pressure on the body surface.

where

E = elastance
R = resistance
V = volume
total PEEP = total PEEP ($PEEP_{set} + PEEP_{auto}$)
dv/dt = the derivative of volume with respect to time

Note that this is a differential equation of a particular type. It is called a first order, linear differential equation with constant coefficients: $f(x) = a \cdot y + b \cdot dy/dx$. In this case, the coefficients, or parameters of the equation (a and b), are elastance and resistance.

Equation 3 is more commonly written using compliance (C) instead of elastance and $\dot{V}$ instead of dv/dt:

$$P_{aw}(t) = \frac{V(t)}{C} + R \cdot \dot{V}(t) + PEEP_{tot} \tag{4}$$

Equation 4 applies to inspiration. For expiration, we assume that P_{aw} is decreased immediately to $PEEP_{set}$ (ie, $P_{aw} = PEEP_{set}$). Then subtracting $PEEP_{set}$ from both sides of Equation 4 yields:

$$0 = \frac{V(t)}{C} + R \cdot \dot{V}(t) + PEEP_{auto} \tag{5}$$

Rearranging results in:

$$\frac{V(t)}{C} + PEEP_{auto} = -R \cdot \dot{V}(t) \tag{6}$$

Table 1. Different meanings of the word "pressure"

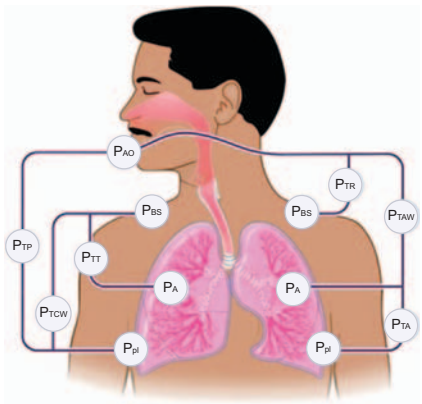
Pressure at a point in space
- Absolute: P_{BS} *(atmospheric pressure)*
- Gauge: P_{aw} *(above atmospheric pressure)*
- Relative: P_{vent}, *pressure support level (above PEEP)*

Pressure difference between 2 points in space
- $P_{aw} - P_{BS} = P_{TR}$ *trans-respiratory pressure*
- $P_{aw} - P_{es} = P_{TP}$ *trans-pulmonary pressure*

Pressure difference between 2 points in time
- $P_{aw}(t_2) - P_{aw}(t_1) = \Delta P_{aw}$
- *Examples:* $P_{0.1}$ *or* P_{occ}

Pressure difference between 2 points in space *and* time
- $P_{TR}(t_2) - P^{TR}(t_1) = \Delta P^{TR}$
- *Examples:* P_{plat}

P_{aw}, pressure at the airway opening; P_{BS}, pressure on body surface; P_{es}, esophageal pressure; P_{plat}, plateau pressure; P_{TR}, trans-respiratory pressure; P_{occ}, airway occlusion pressure.



Symbol	Name
P_{AO}	Airway opening pressure (aka Paw)
P_A	Alveolar pressure
P_{BS}	Body surface pressure
P_{pl}	Pleural pressure

Symbol	Name	Definition
P_{TAW}	Trans-airway pressure difference	$P_{AO} - P_A$
P_{TA}	Trans-alveolar pressure difference	$P_A - P_{pl}$
P_{TCW}	Trans-chest wall pressure difference	$P_{pl} - P_{BS}$
P_{TP}	Trans-pulmonary pressure difference	$P_{AO} - P_{pl}$
P_{TR}	Trans-respiratory pressure difference	$P_{AO} - P_{BS}$
P_{TT}	Trans-thoracic pressure difference	$P_A - P_{BS}$

Fig. 10. Pressures as points in space. There are many points of interest for measuring pressure in the respiratory system. Here are the most common points used in pulmonary mechanics.

Equation 6 shows that the pressure driving expiratory flow (indicated by the negative sign on the right-hand side of the equation) is the elastic pressure stored in the respiratory system associated with the tidal volume and the trapped gas causing $PEEP_{auto}$.

The parameters of resistance and compliance can be derived from ventilator data in 2 ways:

- *Static conditions:* Manual measurements of pressure, volume, and flow changes allow calculation of respiratory system resistance (change in trans-respiratory pressure/flow) and elastance/compliance (change in trans-respiratory pressure/volume) as shown in Figures 12 and 13.
- *Dynamic conditions:* Ventilators commonly use linear regression to fit data points to the EoM, as shown in Figure 14.

Equations for Pressure, Volume, and Flow

From the EoM, we can derive expressions for pressure, volume, and flow under various conditions during mechanical ventilation. The simplest case is ventilation with volume control using a constant inspiratory flow (Supplementary Fig. S1).

The solution for pressure control with constant inspiratory pressure is more complicated. We begin by specifying the constant driving pressure, ΔP, as the inspiratory pressure target above $PEEP_{set,}$ assuming no $PEEP_{auto}$. Then, using the tools for solving differential equations, we find that the expressions for passive inspiratory and expiratory flow and volume are *exponential equations* (Supplementary Fig. S2).

AutoPEEP

For any mode, during passive expiration, $PEEP_{auto}$ is due to any portion of the V_T not exhaled at the end of the expiratory time (ie, trapped gas). Hence, the pressure in the alveolar region of the lungs, P_A, depends on the volume of trapped gas and compliance:

$$P_A = \frac{V_T}{C}\left(e^{-t/RC}\right) \tag{7}$$

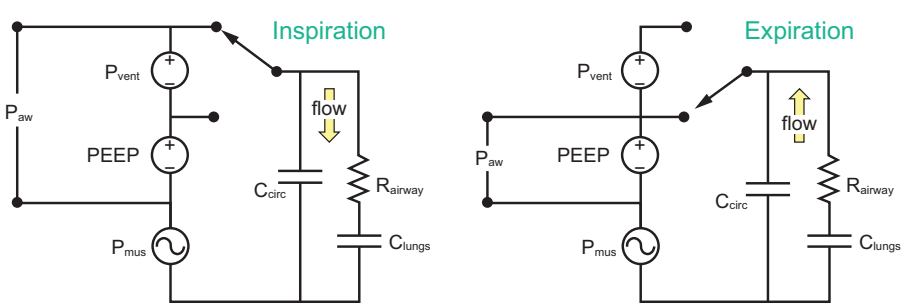


Fig. 11. Electrical circuit components representing a patient–ventilator system during pressure control ventilation.

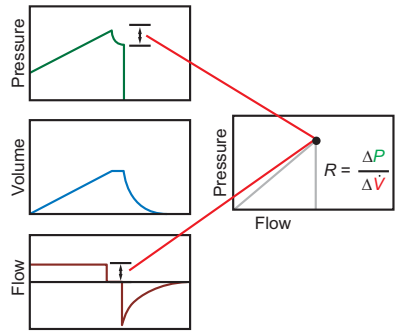


Fig. 12. Calculation of resistance from static ventilator data.

where t is time during expiration. Thus, $PEEP_{tot}$ is the P_A when t = expiratory time (T_E):

$$PEEP_{tot} = \frac{V_T}{C}\left(e^{-T_E/RC}\right) \tag{8}$$

and by definition:

$$PEEP_{auto} = PEEP_{tot} - PEEP_{set} \tag{9}$$

Derivation of the Time Constant

If we plot the exponential equations for pressure, volume, and flow, we get graphs that look like Figure 15. The rate of change of these variables is often expressed in terms of something called the *time constant*. This is a value calculated as the product of resistance and compliance, and it has units of time, usually seconds. In the period equal to one time constant, an exponential function changes in value by 63%. This is relevant to ventilator settings and waveform interpretation because it takes approximately 3–5 time constants for adequate inspiration or expiration.

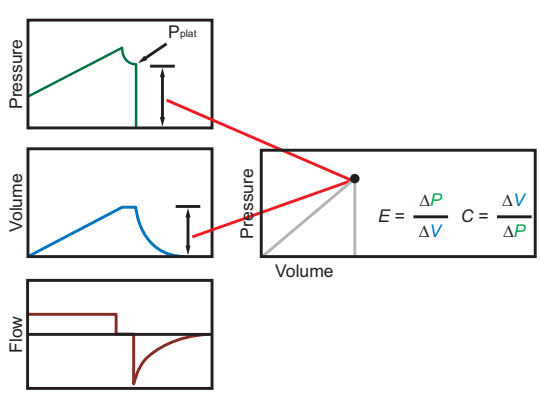


Fig. 13. Calculation of compliance from static ventilator data. Note that the baseline pressure is $PEEP_{tot}$.

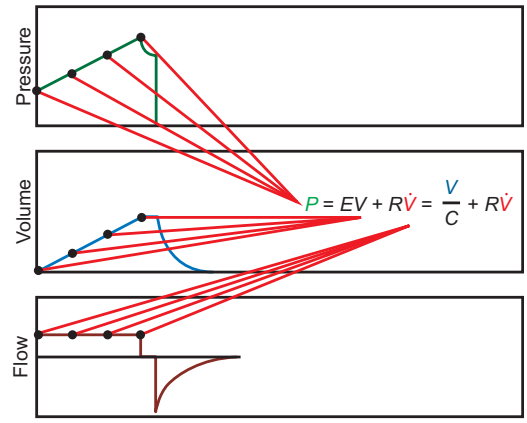


Fig. 14. Calculation of compliance and resistance simultaneously with linear regression using dynamic ventilator data.

Here is something that is virtually never seen in any medical textbook: It is the derivation of the time constant from the EoM. The exponential equations for pressure, volume, and flow have the term RC, or resistance times compliance (eg, Equation 7). If we set inspiratory time equal to RC, then the expression simplifies to e to the power of minus 1. Thus, the whole expression simplifies to the value of 0.63, which is interpreted as a change of 63% of the final inspiratory volume in the period of one time constant. For expiration, a 63% change leaves us with 37% of the steady-state value of zero flow or volume (Supplementary Fig. S3).

Derivations for Unassisted Breathing

We can also find solutions for volume and flow for *unassisted breathing*. A simplifying assumption is that P_{mus} can be adequately represented by a sinusoidal function (Supplementary Fig. S4). The phase angle between sinusoidal functions represents the difference in their starting points (or other reference positions within their cycles) measured in degrees or radians. It indicates how much one sinusoidal wave is shifted relative to another. It is a metric of how "out of sync" they are. A phase angle of 0° or 0 radians means the waves are in phase, whereas a phase angle of 180° or π radians indicates they are maximally out of phase.

It turns out that the solution for volume is also a sinusoidal function that is out of sync with the P_{mus} function by a phase angle that is determined by both the time constant and the breathing frequency (Supplementary Fig. S5).

In the same way, the flow solution is again a sinusoidal function, but with a different phase angle relative to

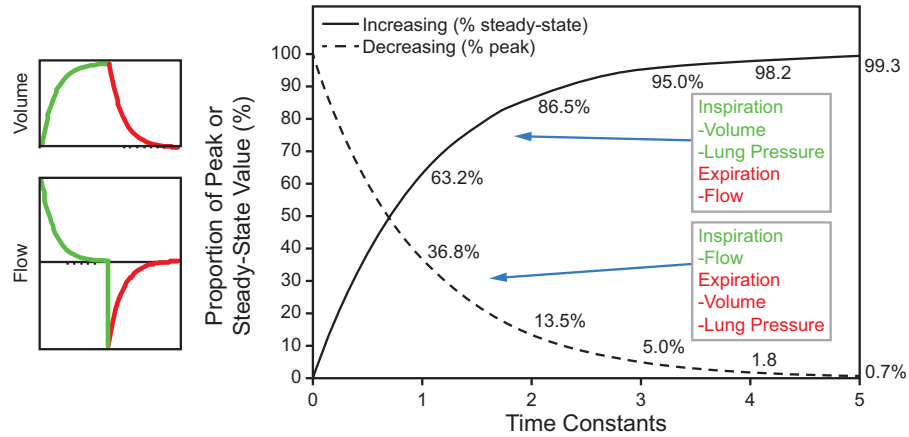


Fig. 15. The time constant describes the time course of an exponential curve.

P_{mus} that is determined by the time constant and the breathing frequency (Supplementary Fig. S6).

Differences in the synchrony between pressure, volume, and flow are easier to appreciate with a graph like the one shown in Figure 16. The peak values are used to determine the relative lag and lead times. In general, flow leads pressure, and pressure leads volume.

Figure 17 shows how the equations derived from the EoM can be used to illustrate pre-set pressure, volume, and flow waveforms generated by ventilators.

Applications of the EoM for Teaching Purposes

If you think that all these equations are way too complicated to be useful at the bedside, then there are some practical simplifications. Start with the EoM in the form shown as Equation 3. First, assume the reader knows that pressure, volume, and flow are functions of time, so we can remove t:

$$P_{aw} = E{\cdot}V + R{\cdot}\dot{V} + PEEP_{set} + PEEP_{auto} \quad (10)$$

Next, subtract $PEEP_{set}$ from both sides of the equation.

$$P_{aw} - PEEP_{set} = E{\cdot}V + R{\cdot}\dot{V} + PEEP_{auto} \quad (11)$$

This subtraction allows us to define P_{vent} as the pressure generated by the ventilator to cause inspiration against the elastic and resistive loads and any $PEEP_{auto}$:

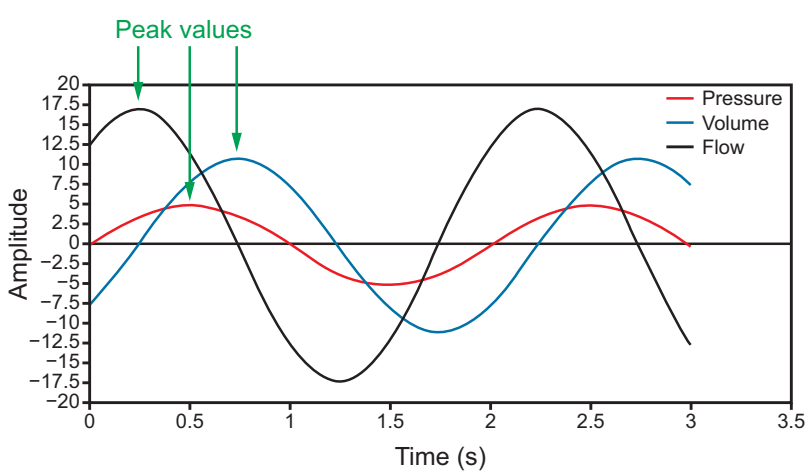


Fig. 16. Graph showing the concept of phase angles between sinusoidal waveforms. The peak value for flow comes before (leads) the peak value for pressure. The peak value for pressure comes before (leads) the peak value for volume.

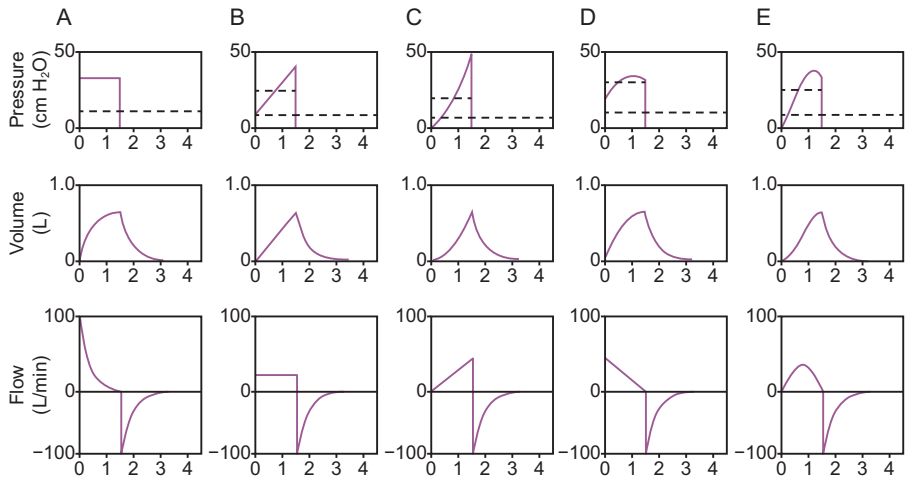


Fig. 17. Idealized pressure, volume, and flow waveforms derived from the EoM for the respiratory system. **(A)** Pressure control with a rectangular pressure waveform. The short-dotted lines represent mean *inspiratory* pressure, and the long-dotted lines represent mean *airway* pressure. Note that for the rectangular pressure waveform, the mean inspiratory pressure is the same as the peak inspiratory pressure. **(B)** Volume control with a constant flow waveform. **(C)** Volume control with an ascending ramp flow waveform. **(D)** Volume control with a descending ramp flow waveform. **(E)** Volume control with a sinusoidal flow waveform.

$$P_{vent} = E{\cdot}V \ + \ R{\cdot}\dot{V} \ + \ PEEP_{auto} \tag{12}$$

Finally, we add P_{mus} as a forcing function to P_{vent} on the left-hand side of the equation:

$$P_{vent} + P_{mus} = E{\cdot}V \ + \ R{\cdot}\dot{V} \ + \ PEEP_{auto} \tag{13}$$

With this expanded simple equation, we can start to define some terms that are essential for understanding patient–ventilator interactions. First, the pressure generated by the volume delivered into the elastic compartment is called the *elastic load*. The pressure required to generate inspiratory flow is called the *resistive load*. If there is any $PEEP_{auto}$, this can be interpreted as an additional elastic load, equivalent to the volume of trapped gas times the respiratory system elastance. These concepts are shown in Figure 18.

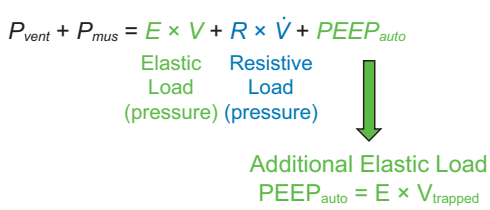


Fig. 18. Definitions of elastic and resistive loads.

One utility of this equation is that we can demonstrate that the total pressure generated by the ventilator and the patient's muscles must exceed the level of $PEEP_{auto}$ to start inspiratory flow and deliver a tidal volume.

$$P_{vent} \ + \ P_{mus} - PEEP_{auto} = E{\cdot}V \ + \ R{\cdot}\dot{V} \tag{14}$$

In other words, the quantity ($P_{vent} + P_{mus}$) minus $PEEP_{auto}$ must be greater than zero for there to be positive flow and volume change (ie, inspiration).

Physical simulators based on the EoM are used to model the respiratory system for simulation purposes (eg, for prototype design and testing required by the United States Food and Drug Administration before marketing). These range from simple test lungs (fixed resistance and compliance) to sophisticated breathing simulators (adjustable resistance, compliance, and P_{mus}) that even have physiological outputs like hemodynamics and gas exchange. A simple passive patient simulator suitable for ventilator performance testing can be constructed from a rigid-walled container (Fig. 19). This simulator can be simplified by removing the pneumotachometer because alveolar volume is simply alveolar pressure times simulator compliance (assuming isothermal conditions are established by the copper wool acting as a thermal damper).[21]

The Taxonomy for Modes of Mechanical Ventilation

We can define a "mode" of ventilation as a predetermined interaction between ventilator and patient.[22] The

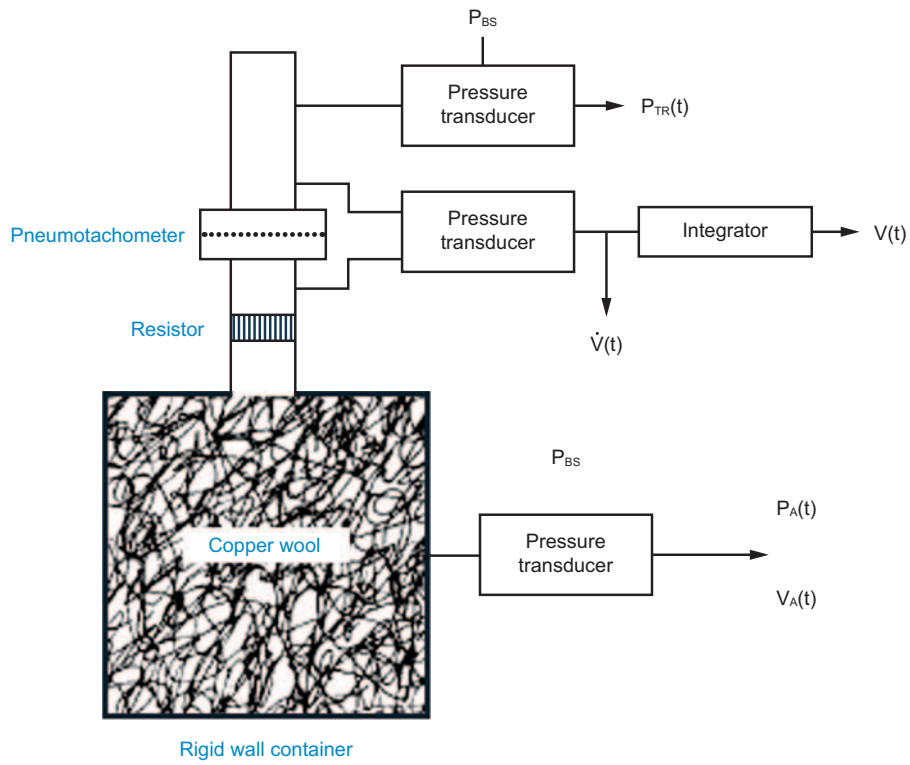


Fig. 19. Passive lung simulator suitable for ventilator performance testing constructed from a rigid-walled container. P_A, alveolar pressure; P_{TR}, trans-respiratory pressure; V_A, alveolar volume; $\dot{V}$, flow.

variety of modes currently available has exploded over the last several decades. This proliferation demands the use of a classification system, or taxonomy, for the same reasons we use generic drug names.[22–24] For example, the most recent textbook on respiratory care equipment lists 495 unique mode brand names on 55 different ventilators, which can be reduced to 74 unique mode TAGs (taxonomic attribute groupings) using the mode taxonomy.[25] At the time of this writing, the free app called Ventilator Mode Map (available for iPhone and Android) contains 569 unique mode brand names on 78 ventilators, which are classified using 94 unique classification TAGs (as classified by the author, who designed the app). Individual ventilators may have several dozen different modes. Is there any wonder that clinicians are confused about the "tools in the toolbox"? The mode taxonomy has been adopted virtually verbatim in a wide variety of current textbooks and has been referenced in a growing number of articles published in peer-reviewed medical journals (Supplementary Table S1).

In addition to providing a basis for understanding ventilators, the taxonomy is important for ordering modes and charting patient outcomes in the medical record.[26] Also, quality improvement through data mining of electronic medical records is severely hampered when random terms are used for modes.

Brief Introduction to Ventilator Waveform Analysis

All the above information can be brought to bear on the task of ventilator waveform analysis. To establish a foundational understanding, we must memorize the idealized waveforms associated with passive volume control and pressure control, both with set-point targeting. These are specified by the EoM, where inspiratory and expiratory R are assumed to be equal. Any deviations from these that you observe for actual patients will give clues as to the status of patient–ventilator interactions.

Reference Waveforms for Normal Mechanics

Figures 20 and 21 show idealized pressure, flow, and volume waveforms for normal respiratory system mechanics for an intubated adult (r = 10 cm H_2O, C = 60 mL/cm H_2O) under passive conditions (P_{mus} = 0 cm H_2O) for continuous mandatory ventilation (CMV) and continuous spontaneous ventilation (CSV) modes (respectively). The targeting schemes are all set-point. Note that the waveforms are presented in a specific order (ie, pressure, flow, volume) because volume generally gives the least amount of information.

Pressure control with active breathing and adaptive targeting is more challenging to identify, but can be recognized by the fact that inspiratory pressure decreases as inspiratory effort increases (Fig. 22).

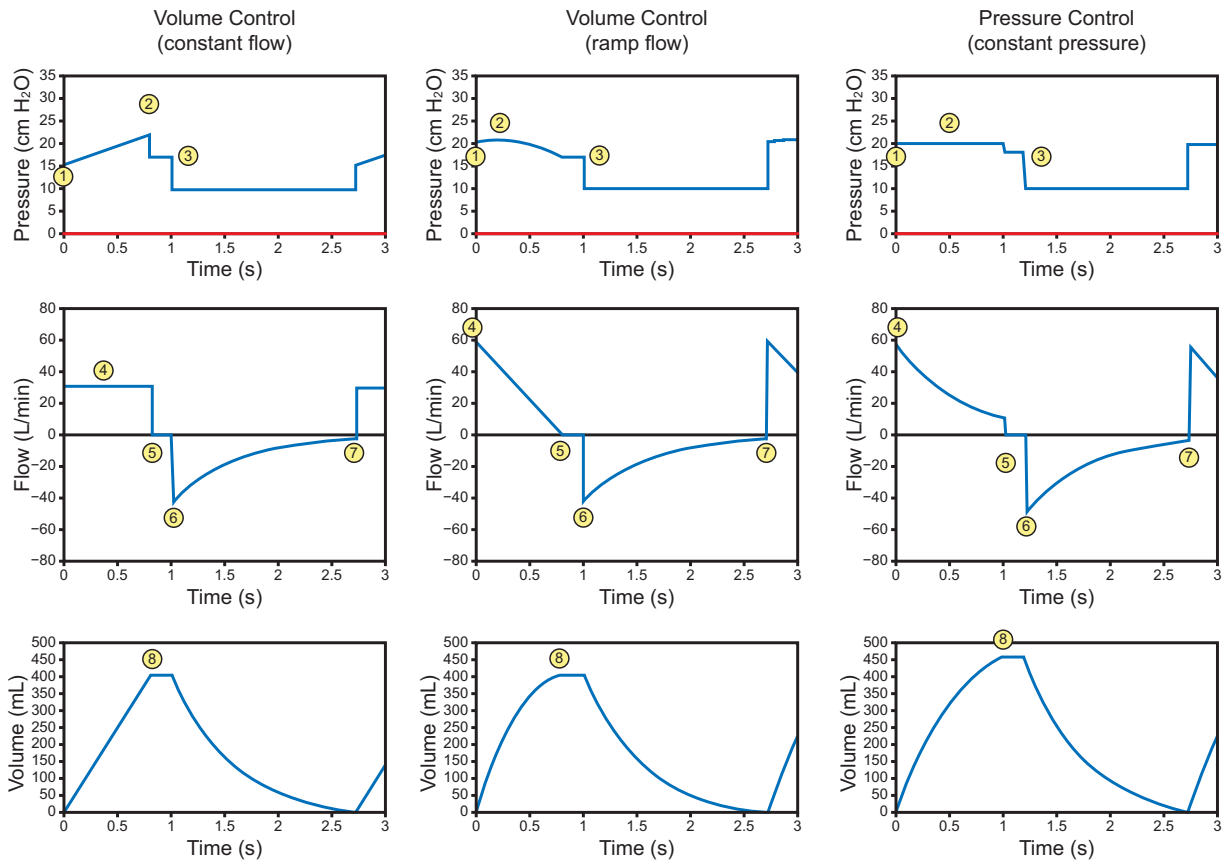


Fig. 20. Idealized reference waveforms for passive volume control and pressure control CMV modes using set-point targeting. Simulation parameters: R = 10 cm H_2O/L/s, C = 60 mL/cm H_2O, P_{mus} = 0 cm H_2O. 1 = initial pressure rise (relative to PEEP), 2 = peak inspiratory pressure, 3 = plateau pressure (relative to atmospheric pressure), 4 = peak inspiratory flow, 5 = end-inspiratory flow (ie, at end of inspiratory flow time), 6 = peak expiratory flow, 7 = end-expiratory flow (ie, at end of expiratory flow time), 8 = peak volume or tidal volume (corresponds in time to end-inspiratory flow).

Similarly, pressure control with servo targeting is identified by the fact that inspiratory pressure is proportional to inspiratory effort (Fig. 23).

Procedure for Interpreting Ventilator Waveforms

The key skill for ventilator waveform analysis is knowing where to look for deviations from ideal waveform shapes, again dictated by the EoM in this form:

$$P_{vent} + P_{mus} = E \cdot V + R \cdot \dot{V} + PEEP_{auto} \qquad (15)$$

See also the equations in Supplementary Data.

Rules of thumb

1. For volume control (VC), changes in mechanics affect airway pressure because volume and flow are pre-set.
2. For pressure control (PC), changes in mechanics affect the flow and volume waveforms.
3. For a single-compartment model represented by Equation 15, inspiratory and expiratory values for E and R are the same. In real life, E is virtually always the same on inspiration and expiration, but R may be higher (eg, in the case of flow limitation because of collapsible airways). In this case, the effect of low expiratory flow may mask the effect of high elastance (ie, if R is high and C is low, then the time constant may not be different from normal).
4. Waveform distortions because of P_{mus} occurring at random times during the breath may make the identification of some or all of these reference points impossible.

The following are waveform reference points (see numbers in Figs. 20 and 21) and associated interpretation

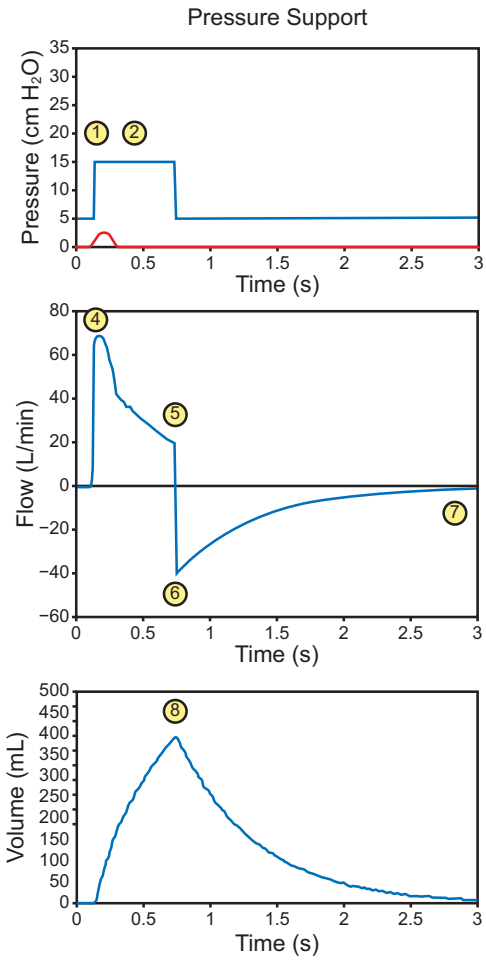


Fig. 21. Idealized reference waveforms for active pressure control continuous spontaneous ventilation (PC-CSV) using set-point targeting (pressure support). Simulation parameters: R = 10 cm H_2O/L/s, C = 60 mL/cm H_2O, P_{mus} = 2.5 cm H_2O. Numbers have the same meanings as Figure 20 (note there is no number 3 because there is no inspiratory hold and no P_{plat}). P_{plat}, plateau pressure.

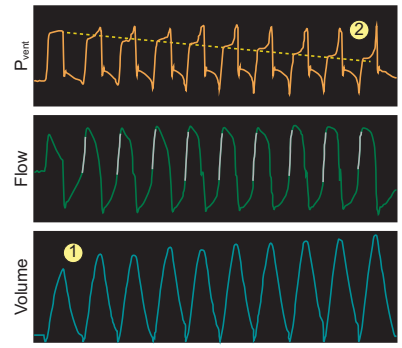


Fig. 22. Pressure control with adaptive targeting (eg, mode called pressure regulated volume control). (1) Increasing P_{mus} results in increasing tidal volume delivery beyond the set target indicated by dotted line. (2) In response, the ventilator automatically decreases the inspiratory pressure target indicated by the dotted line. The sharp increase in airway pressure at the end of the pre-set inspiratory time is because of P_{mus} relaxation. P_{mus}, muscle pressure.

rationales. Remember that these apply most closely to idealized waveforms (eg, generated by a mathematical model). They may not all be identifiable in real-world waveforms that are distorted by a variety of factors (eg, patient circuit mechanics and ventilator control system delays).

1. Initial pressure rise above initial PEEP
   - *VC:* Owing to the term $R \cdot \dot{V}$ in the EoM, so that any change in R will cause a proportional change in the initial pressure rise. This pressure is increased (ie, larger than PIP – P_{plat}) if there is airway closure and in this context has been called "conductive pressure."[27]
   - *PC:* P_{init} is usually identical to set peak inspiratory pressure by design (for set-point targeting scheme) and independent of R and C. Ideally, the initial pressure rise is instantaneous because the flow rise is instantaneous. However, some ventilators offer a setting called something like "pressure rise time" that can decrease the slope of the pressure rise, with the result that the peak inspiratory flow is decreased.
2. Peak inspiratory pressure ($P_{peak\text{-}insp}$ or PIP)
   - *VC:* Inspiratory pressure is the sum of the terms V/C and term $R \cdot \dot{V}$ in the EoM so that any change in C or R will cause a corresponding change in peak pressure (ie, peak pressure decreases with increasing C but increases with increasing R). *Note: For VC with constant flow, the slope of the line connecting points 1 and 2 is proportional to elastance (ie, the higher the slope, the higher the elastance and lower the compliance).*
   - *PC:* PIP is pre-set (for set-point targeting) and independent of R and C.
3. Plateau pressure (P_{plat})
   - *VC:* P_{plat} is measured relative to atmospheric pressure (ie, body surface pressure, P_{BS}) and is the sum of the terms V/C and $PEEP_{tot}$ in the EoM. The difference between P_{peak} and P_{plat} for

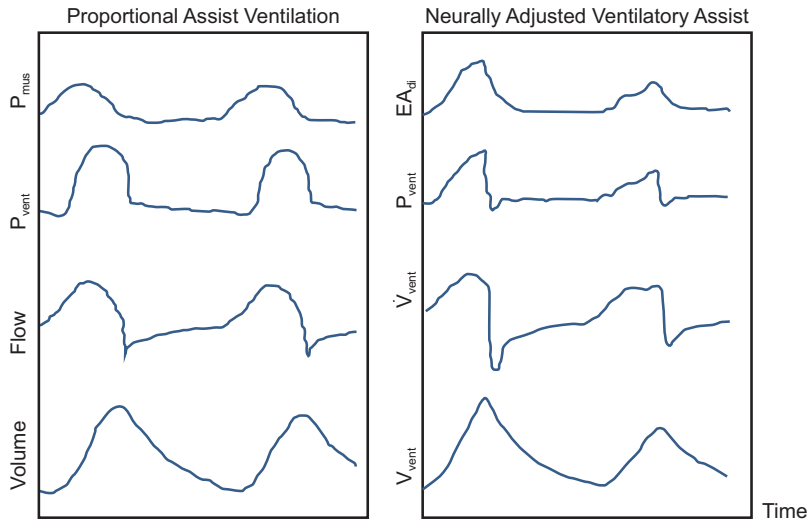


Fig. 23. Pressure control with servo targeting (eg, modes called proportional assist and neurally adjusted ventilatory assist). P_{vent}, pressure delivered by ventilator; $\dot{V}_{vent}$, flow delivered by ventilator; V_{vent}, volume delivered by ventilator.

VC with constant flow is because of the term $R \cdot \dot{V}$ in the EoM, so that increases in this value (for the same flow) will indicate increases in R. Distortions in the expected (passive) pressure waveform between the initial pressure rise and P_{peak} indicate the effects of P_{mus}.

- *PC:* P_{plat} is measured relative to body surface pressure (ie, P_{BS}) and is the sum of the terms V/C and $PEEP_{tot}$ in the EoM. For PC, P_{plat} is identical with set PIP if flow at end inspiration is zero and the patient is passive. However, if flow is not zero at end inspiration for a passive patient, then the full tidal volume has not been delivered, and P_{plat} will be less than the set PIP. Finally, if P_{mus} during inspiration contributes to a larger V_T than passive conditions, then P_{plat} will be higher than the set PIP.
- *VC and PC:* Distortions in the expected (passive) pressure waveform during the inspiratory hold indicate the effects of P_{mus}.

4. Peak inspiratory flow ($\dot{V}_{peak-insp}$)
   - *VC:* Pre-set and independent of R and C

Table 2. Increased elastic load in CMV

Step	*Volume control CMV (constant or ramp flow)*	*Pressure control CMV (constant pressure)*
1	P_{init}: Unchanged	P_{init}: Unchanged (pre-set)
2	$P_{peak-insp}$: Increased because of increased elastic load (E·V), offset slightly by decreased $PEEP_{auto}$ indicated by decreased end-expiratory flow	$P_{peak-insp}$: Unchanged (pre-set)
3	P_{plat}: Increased because of increased elastic load (E·V), offset slightly by decreased $PEEP_{auto}$ as indicated by decreased end-expiratory flow	P_{plat}: Increased because of increased elastic load (E·V), offset slightly by decreased $PEEP_{auto}$ indicated by decreased end-expiratory flow
4	$\dot{V}_{peak-insp}$: Unchanged (pre-set)	$\dot{V}_{peak-insp}$: Slightly increased because of slightly increased ΔP. Recall that $\dot{V} = \Delta P/R$. The driving pressure for inspiratory flow ($\Delta P = PEEP_{set} - PEEP_{tot}$) is slightly larger because of a slightly lower $PEEP_{auto}$
5	$\dot{V}_{end-insp}$: Unchanged (pre-set)	$\dot{V}_{end-insp}$: Decreased because of decreased time constant ($\tau = R \cdot C = R/E$)
6	$\dot{V}_{peak-exp}$: Increased because of increased P_{plat}. Recall that $\dot{V} = -\Delta P/R$ and that the −ΔP for expiratory flow is $PEEP_{set} - P_{plat}$	$\dot{V}_{peak-exp}$: Increased because of increased Pplat. Recall that $\dot{V} = -\Delta P/R$ and that the −ΔP for expiratory flow is $PEEP_{set} - P_{plat}$
7	$\dot{V}_{end-exp}$: Decreased because of decreased time constant ($\tau = R \cdot C = R/E$)	$\dot{V}_{end-exp}$: Decreased because of decreased time constant ($\tau = R \cdot C = R/E$)
8	V_T: Unchanged (pre-set)	V_T: Decreased because of increased E. Recall that maximum $V_T = \Delta/E = \Delta C$

$P_{peak-insp}$, peak inspiratory pressure; P_{init}, initial pressure rise above PEEP; $\dot{V}_{end-exp}$, end-expiratory flow; $\dot{V}_{peak-exp}$, peak expiratory flow.

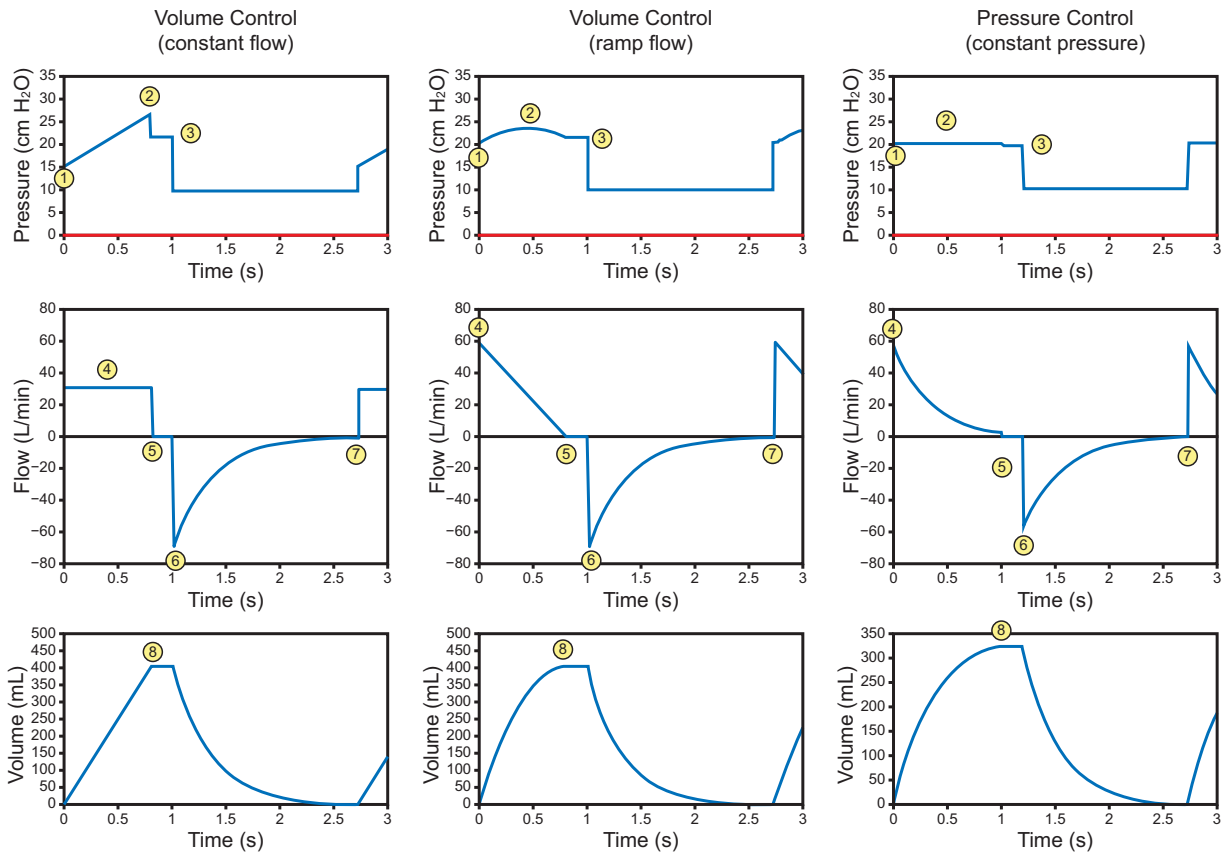


Fig. 24. Increased elastic load. Idealized waveforms for passive volume control and pressure control CMV modes using set-point targeting. Simulation parameters: R = 10 cm H_2O/L/s, C = 35 mL/cm H_2O, P_{mus} = 0 cm H_2O.

- *PC:* Peak inspiratory flow is equal to ΔP/R, where ΔP = PIP − $PEEP_{tot}$, so that increases in $PEEP_{auto}$ or R will decrease peak flow

5. End-inspiratory flow ($\dot{V}_{end-insp}$)
   - *VC:* When the pre-set V_T is delivered, end-inspiratory flow drops to zero. This corresponds to the end of the inspiratory flow time (time during which flow is positive), which is the

Table 3. Increased resistive load in CMV

Step	*Volume control CMV (constant or ramp flow)*	*Pressure control CMV (constant pressure)*
1	P_{init}: Increased because of increased resistive load (R·$\dot{V}$), augmented slightly by increased $PEEP_{auto}$ indicated by increased end-expiratory flow	P_{init}: Unchanged (pre-set)
2	$P_{peak\text{-}insp}$: Increased because of increased resistive load (R·$\dot{V}$), augmented slightly by increased $PEEP_{auto}$ as indicated by increased end-expiratory flow	$P_{peak\text{-}insp}$: Unchanged (pre-set)
3	P_{plat}: Increased slightly because of increased $PEEP_{auto}$ as indicated by increased end-expiratory flow	P_{plat}: Decreased because of decreased V_T
4	$\dot{V}_{peak-insp}$: Unchanged (pre-set)	$\dot{V}_{peak-insp}$: Decreased because of increased R. Recall that peak inspiratory flow = (P_{peak} − $PEEP_{tot}$)/R
5	$\dot{V}_{end-insp}$: Unchanged (pre-set)	$\dot{V}_{end-insp}$: Increased because of increased time constant (τ = R·C = R/E), offset slightly by decreased V_T
6	$\dot{V}_{peak-exp}$: Decreased because of increased R. Recall that $\dot{V}$ = −ΔP/R and that the −ΔP for expiratory flow is $PEEP_{set}$ − P_{plat}	$\dot{V}_{peak-exp}$: Decreased because of increased R. Recall that $\dot{V}$ = −ΔP/R and that the −ΔP for expiratory flow is $PEEP_{set}$ − P_{plat}, further decreased by decreased P_{plat}
7	$\dot{V}_{end-exp}$: Increased because of decreased time constant (τ = R·C = R/E)	$\dot{V}_{end-exp}$: Increased because of increased time constant (τ = R·C = R/E), offset slightly by decreased V_T
8	V_T: Unchanged (pre-set)	V_T: Decreased because of increased inspiratory time constant (τ = R·C = R/E)

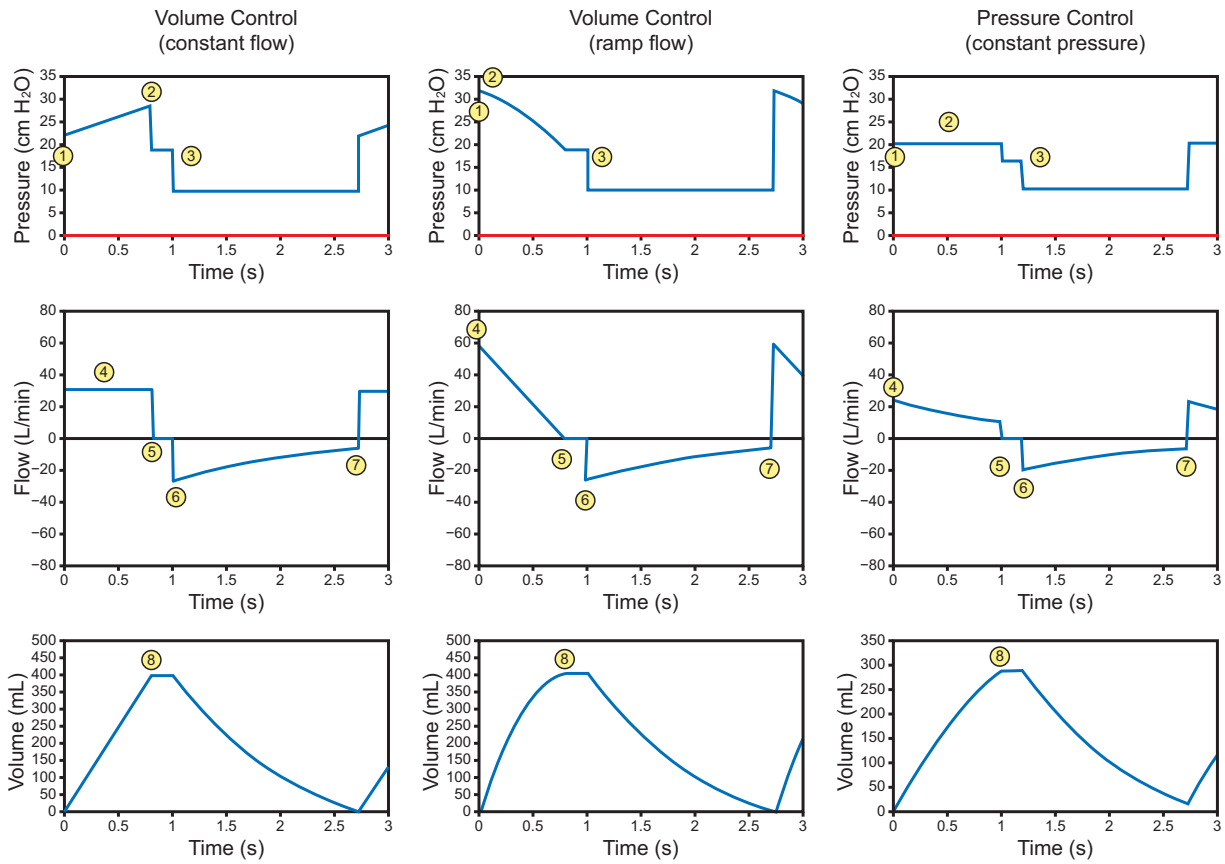


Fig. 25. Increased resistive load. Idealized waveforms for passive volume control and pressure control CMV modes using set-point targeting. Simulation parameters: R = 20 cm H_2O/L/s, C = 60 mL/cm H_2O, P_{mus} = 0 cm H_2O.

same as inspiratory time unless an inspiratory hold (period of airway occlusion) has been set.

- *PC:* Flow at end inspiration may be zero, positive, or negative, depending on the design of the ventilator and any effects of P_{mus}. For a passive patient, end-inspiratory flow corresponds to the end of the inspiratory flow time, which is the same as inspiratory time unless an inspiratory hold (period of airway occlusion) has been set. The value of flow at this time is determined by the inspiratory time

Table 4. Increased inspiratory effort (P_{mus}) in CMV

Step	*Volume control CMV (constant or ramp flow)*	*Pressure control CMV (constant pressure)*
1	P_{init}: Unchanged in this simulation—in reality, if the breath had been triggered by P_{mus} then this would be decreased	P_{init}: Unchanged (pre-set)
2	$P_{peak\text{-}insp}$: *Constant flow*—unchanged because effort ended before delivery of V_T. *Ramp Flow*—decreased because P_{mus} occurred during the time when peak pressure would have occurred under passive conditions	$P_{peak\text{-}insp}$: Unchanged (pre-set)
3	P_{plat}: Unchanged because effort ended before start of inspiratory hold	P_{plat}: Increased because of increased V_T
4	$\dot{V}_{peak-insp}$: Unchanged (pre-set)	$\dot{V}_{peak-insp}$: Increased because P_{mus} adds to P_{vent} early in the inspiratory time. Flow is increased for every instant that P_{mus} is higher than zero
5	$\dot{V}_{end-insp}$: Unchanged (pre-set)	$\dot{V}_{end-insp}$: Unchanged in this simulation but depends on magnitude and duration of P_{mus}
6	$\dot{V}_{peak-exp}$: Unchanged because effort does not extend into the expiratory time	$\dot{V}_{peak-exp}$: Increased because of increased P_{plat}. Recall that $\dot{V} = -\Delta P/R$ and that the $-\Delta P$ for expiratory flow is $PEEP_{set} - P_{plat}$
7	$\dot{V}_{end-exp}$: Unchanged because effort does not extend into the expiratory time	$\dot{V}_{end-exp}$: Increased because of increased V_T
8	V_T: Unchanged (pre-set)	V_T: Increased because P_{mus} adds to P_{vent}, increasing the area under the flow curve

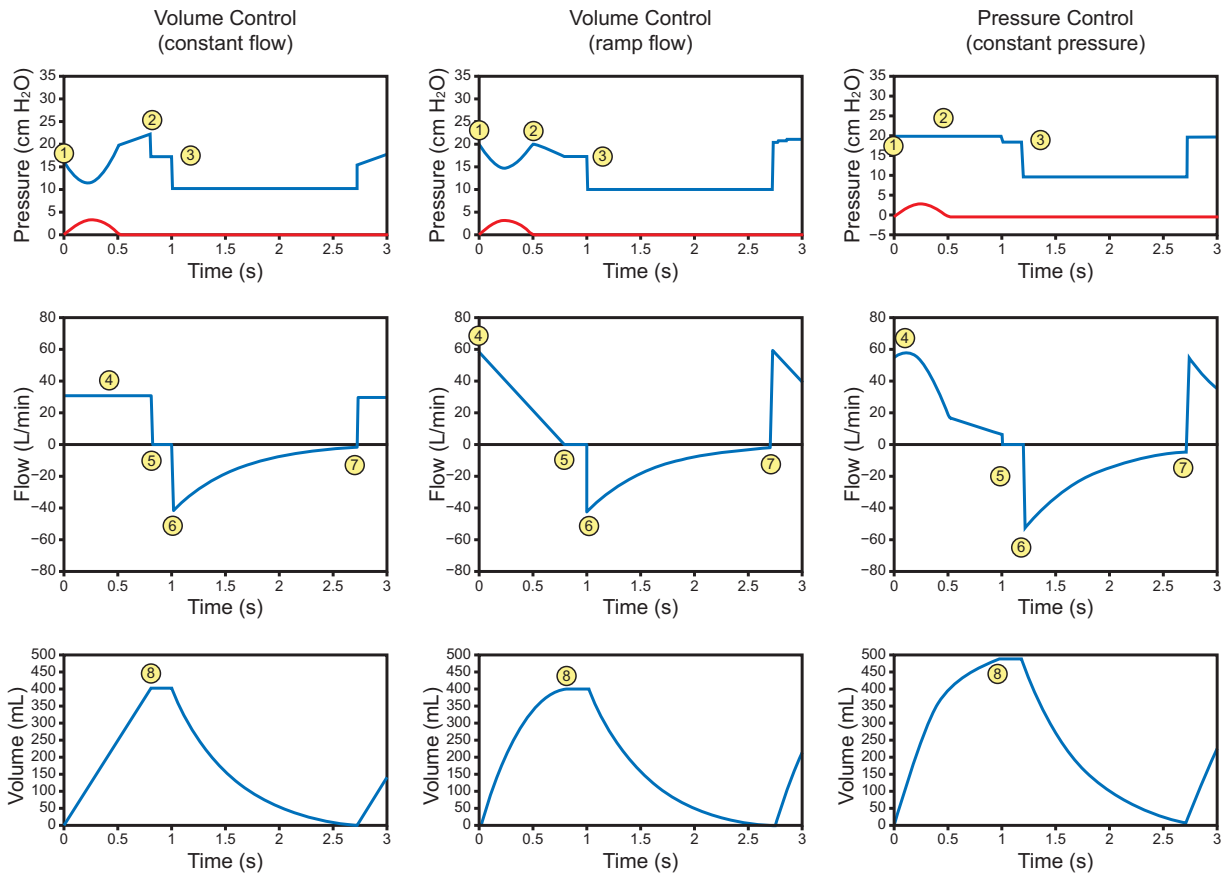


Fig. 26. Increased inspiratory effort. Idealized waveforms for passive volume control and pressure control CMV modes using set-point targeting. Simulation parameters: R = 10 cm H_2O/L/s, C = 60 mL/cm H_2O, P_{mus} = 3 cm H_2O.

constant and the inspiratory flow time. End-inspiratory flow for flow-cycled breaths (e.g., pressure support) is usually expressed as a percentage of peak inspiratory flow. Distortions in the expected (passive) inspiratory flow waveform during inflation indicate the effects of P_{mus}. *Note: For passive inspiration, the time course of the exponential increase in flow between points 4 and 5 is determined by the time constant (ie, the longer the time constant, the slower the flow during this period). Inspiration is 95% complete after 3 time constants.*

6. Peak expiratory flow ($\dot{V}_{peak-exp}$)
   - *VC and PC:* Peak expiratory flow is equal to $-\Delta P/R$, where $\Delta P = P_{set} - P_{plat}$. The effect of a change in $PEEP_{set}$ on peak expiratory flow depends on how the change affects P_{plat}.
7. End-expiratory flow ($\dot{V}_{end-exp}$)
   - *VC and PC:* End-expiratory flow for passive expiration is determined by the expiratory time constant and the expiratory flow time (time during which flow is negative). Distortions in the expected (passive) expiratory flow waveform during inflation indicate the effects of P_{mus}. *Note: For passive expiration, the time course of the exponential decrease in flow between points 6 and 7 is determined by the time constant (ie, the longer the time constant, the slower the flow during this period). Expiration is 95% complete after 3 time constants.*
8. Tidal volume (V_T)
   - *VC:* Pre-set
   - *PC:* The maximum V_T is determined by $\Delta P \times C$ where $\Delta P = P_{peak} - PEEP_{tot}$. The actual delivered V_T is a function of the inspiratory flow time and the inspiratory time constant. Distortions in the expected (passive) volume waveform during inflation indicate the effects of P_{mus}. During the inspiratory time, a positive change in P_{mus} (inspiratory effort) will increase tidal volume, and a negative change (expiratory effort) will decrease tidal

Table 5. Increased loads in CSV

Step	Increased elastic load	Increased resistive load	Increased inspiratory effort (P_{mus})
1	P_{init}: Unchanged (pre-set)	P_{init}: Unchanged (pre-set)	P_{init}: Unchanged (pre-set)
2	$P_{peak\text{-}insp}$: Unchanged (pre-set)	$P_{peak\text{-}insp}$: Unchanged (pre-set)	$P_{peak\text{-}insp}$: Unchanged (pre-set)
3	P_{plat}: not shown (not available on many ventilators)	P_{plat}: not shown (not available on many ventilators)	P_{plat}: not shown (not available on many ventilators)
4	$\dot{V}_{peak-insp}$: Unchanged. Recall that peak inspiratory flow equals $\Delta P/R$, and R is unchanged. The driving pressure, ΔP, for inspiratory flow is (set PIP – $PEEP_{tot}$) + ΔP_{mus} and this is unchanged	$\dot{V}_{peak-insp}$: Decreased. Recall that peak inspiratory flow equals $\Delta P/R$, and R is increased. The driving pressure, ΔP, for inspiratory flow is (set PIP – $PEEP_{tot}$) + ΔP_{mus} and this is unchanged	$\dot{V}_{peak-insp}$: Increased because of increased ΔP_{mus}. Recall that peak inspiratory flow equals $\Delta P/R$, and R is unchanged. The driving pressure, ΔP, for inspiratory flow is (set PIP – $PEEP_{tot}$) + ΔP_{mus} and ΔP_{mus} increased
5	$\dot{V}_{end-insp}$: Unchanged because the pre-set cycle threshold as a percent of peak inspiratory flow is the same, and $\dot{V}_{peak\text{-}insp}$ is unchanged	$\dot{V}_{end-insp}$: Decreased because the pre-set cycle threshold as a percent of peak inspiratory flow is the same but $\dot{V}_{peak\text{-}insp}$ is decreased	$\dot{V}_{end-insp}$: Increased because the pre-set cycle threshold as a percent of peak inspiratory flow is the same but $\dot{V}_{peak-insp}$ is increased
6	$\dot{V}_{peak-exp}$: Slightly increased because of slightly increased P_{plat}. Recall that $\dot{V} = -\Delta P/R$ and that the $-\Delta P$ for expiratory flow is $PEEP_{set} - P_{plat}$. P_{plat} was not measured but can be estimated as $V_T/C + PEEP_{set}$. For the reference waveform, this was about 12 cm H_2O. Here it is 13 cm H_2O	$\dot{V}_{peak-exp}$: Decreased because of increased R. Recall that $\dot{V} = -\Delta P/R$ and that the $-\Delta P$ for expiratory flow is $PEEP_{set} - P_{plat}$. P_{plat} was not measured but can be estimated as $V_T/C + PEEP_{set}$. For the reference waveform, this was about 12 cm H_2O. Here it is also about 12 cm H_2O, so ΔP is unchanged	$\dot{V}_{peak-exp}$: Unchanged. Recall that $\dot{V} = -\Delta P/R$ and that the $-\Delta P$ for expiratory flow is $PEEP_{set} - P_{plat}$. P_{plat} was not measured but can be estimated as $V_T/C + PEEP_{set}$. For the reference waveform, this was about 12 cm H_2O. Here it is also about 12 cm H_2O, so ΔP is unchanged
7	$\dot{V}_{end-exp}$: Decreased because of decreased time constant ($\tau = R{\cdot}C = R/E$)	$\dot{V}_{end-exp}$: Increased because of increased time constant ($\tau = R{\cdot}C = R/E$)	$\dot{V}_{end-exp}$: Decreased slightly because of the larger tidal volume, but the change is virtually imperceptible
8	V_T: Decreased because of decreased inspiratory time. Recall that V_T during pressure control is a function of ΔP (see #4) and inspiratory time. Inspiratory time is not pre-set but is the result of the set flow cycle threshold (as a percent of peak flow) and the inspiratory time constant. The inspiratory time constant ($\tau = R{\cdot}C = R/E$) decreased because of increased E (decreased C). With a decreased time constant, the time to reach the cycle threshold (ie, the inspiratory time) is decreased, and hence inspiratory time increased	V_T: Increased because of increased inspiratory time. Recall that V_T during pressure control is a function of ΔP (see #4) and inspiratory time. Inspiratory time is not pre-set but is the result of the set flow cycle threshold (as a percent of peak flow) and the inspiratory time constant. The inspiratory time constant ($\tau = R{\cdot}C = R/E$) increased because of increased R. With an increased time constant, the time to reach the cycle threshold (ie, the inspiratory time) is increased, and hence inspiratory time increased	V_T: Increased because P_{mus} adds to P_{vent}

volume. *Note: For passive inspiration, the time course of the exponential increase in volume between zero and point 8 is determined by the time constant (ie, the longer the time constant, the lower the volume during this period). Volume delivery is 95% complete after 3 time constants.*

Waveform Distortions Because of Changes in Mechanics for Continuous Mandatory Ventilation

In this section, we examine a series of waveforms that indicate changes in respiratory system mechanics compared with the reference waveforms for CMV (Fig. 20). We will use the procedure from the last section to determine what kind of change occurred. For simplicity, only single changes in R, E, and P_{mus} will be considered.

Increased elastic load

Table 2 shows the results of the waveform analysis, and Figure 24 shows the waveform distortions for an increase in E (decrease in C).

Increased resistive load

Table 3 shows the results of the waveform analysis, and Figure 25 shows the waveform distortions for an increase in R.

Increased inspiratory effort (P_{mus})

Table 4 shows the results of the waveform analysis, and Figure 26 shows the waveform distortions for an increase in P_{mus}.

Waveform Distortions Because of Changes in Mechanics for Continuous Spontaneous Ventilation with Set-Point Targeting

In this section, we examine a series of waveforms that indicate changes in respiratory system mechanics compared with the reference waveforms for CSV (Fig. 21). We will use the same procedure to determine what kind of change occurred. Again, for simplicity, only single changes in R, E, and P_{mus} will be considered. The P_{mus} change is simple and arbitrary, but in real life, it is complicated and variable.

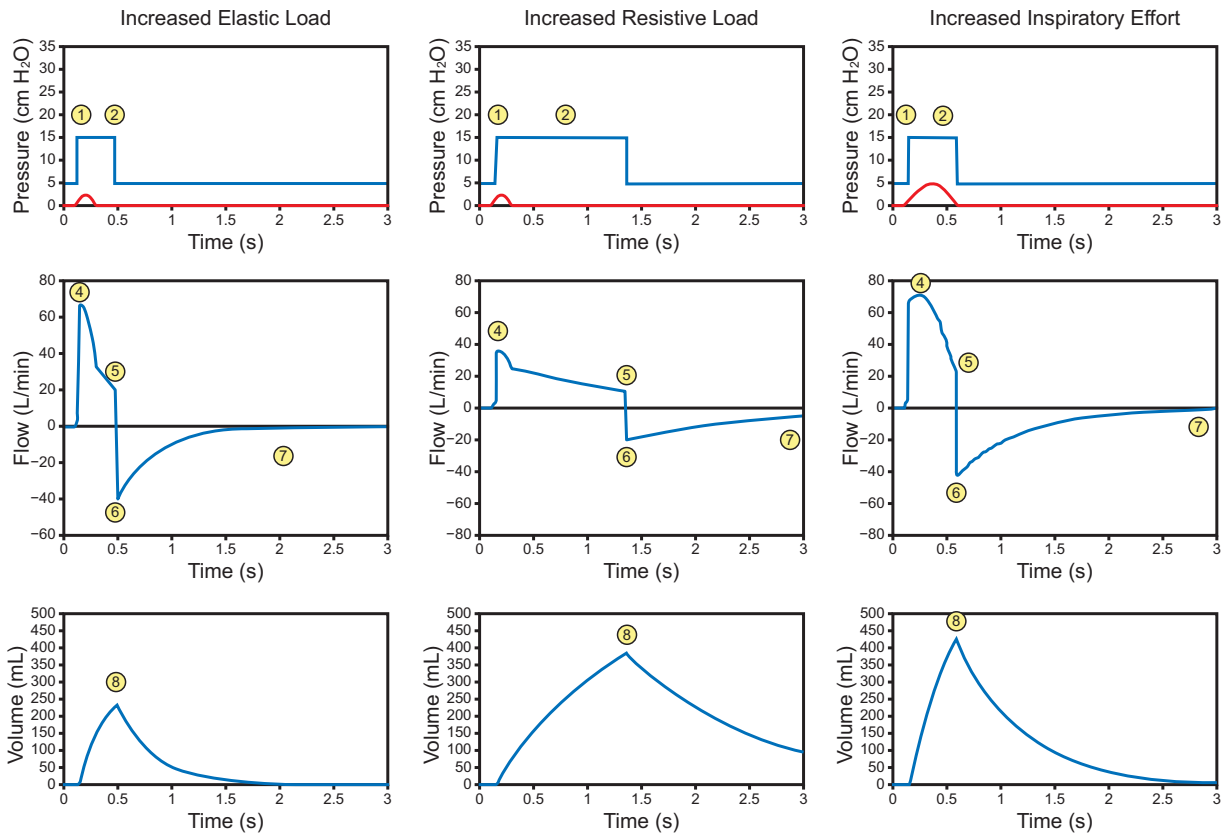


Fig. 27. Increased loads in pressure support.

Table 5 shows the results of the waveform analysis, and Figure 27 shows the waveform distortions of PC-CSVs (pressure support).

Summary

Analogous to reading ECG waveforms, interpretation of ventilator waveforms is an essential skill for any health care provider responsible for patients who are mechanically ventilated. However, this skill is seldom taught in sufficient detail within any educational venue. Two key aspects of waveform interpretation are recognition of the mode of ventilation and identification of any deviations from expected waveshapes because of abnormal loads or breathing efforts. Recognition of the mode relies on the use of the taxonomy for modes of ventilation. Identification of the loads depends on an understanding of the physics and mathematics of lung mechanics and how changes in mechanics are reflected in waveform changes. Both skills depend on at least an intuitive understanding of the EoM for the respiratory system. Mastery of these topics will allow the clinician to recognize and respond to changes in physical patient–ventilator interactions and provide the basis for further study of patient–ventilator discordances.

To gain these skills, we train using idealized pressure, volume, and flow waveforms. These are generated by mathematical models derived from the EoM (eg, programmed into a patient–ventilator simulator) and show none of the real-world artifacts that we normally see on a ventilator display. These include noise (typically oscillation) in pressure and flow waveforms because of unavoidable imperfections of pressure and flow control dampening of the ventilator's targeting scheme. Other artifacts may be because of problems with the patient circuit, like water (condensation) or minor leaks. The point is that if you know what an ideal waveform should look like, then you can identify any deviations and deduce appropriate responses.

Acknowledgments

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Supplementary Data

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Panel Discussion

Owens: I'm thinking about, and you mentioned this a few times, how hard it is teaching these modes to learners in the ICU. It's really tough, right? I don't mean the math and physics, I mean the terminology, all the different versions. I usually teach about assist control and pressure support and I kind of stop there. I guess my question to you is, and to be deliberately provocative, do we really need all these modes?

Chatburn: That's a good question. And the truth is that we don't have evidence. As far as I know in all of the history of research in mechanical ventilation there's only been one mode that's been shown to be substantially better than conventional, and it's high frequency, maybe, for neonates. Will we ever get hard evidence? No way, because if you consider the head-to-head comparisons of all these modes it would take billions and billions of dollars and hundreds of years to do. It's never going to happen. We have to rely on first principles, and we have to rely on whatever evidence we can gain from the animal lab and from simulation and common sense. But it's getting harder and harder to do those kinds of things with this confusion about what you have available, how many tools you have in your toolbox. If all you know is assist control, CPAP, and pressure support, well, you've only got three tools in your toolbox. And when I used to teach residents and fellows about this, I would ask them, 'how many different drugs do you use and how familiar are you with them?' They'd say ten or twenty. 'OK, how many ventilator modes do you use?' One or two. Not too many tools in the toolbox. So what our effort has been at the clinic is to try to provide a standardized way of presenting this complexity of different modes. Why they're different, how you talk about them, how you select them based on how they serve the goals of ventilation. So if you can clinically assess the patient for what the goal is, then you should be able to pick the right tool from the toolbox. A hammer is no good if you want to turn a screw in. You could hammer the screw down and force it in, but clearly there is a better tool.

Mireles-Cabodevila: Rob, can you talk about the technological features that exist for comfort in terms of synchrony and proportionality of the work of breathing? Can you talk a little bit about how that those technological features serve the goal?

Chatburn: Basically what we're looking at are ways that the patient can trigger and cycle the ventilator. First

of all, if the patient has substantial control of the timing of the breath (ie, the trigger and cycle events) then the breath is likely to be more synchronous with the patient's demand than if the ventilator imposes it with a set rate (trigger) or set inspiratory time (cycle). Therefore, if you can have all spontaneous breaths, that's good. If you can have mandatory breaths but only when spontaneous breaths are insufficient to support the minute ventilation, that's better. There are now algorithms that automatically adjust the trigger and cycle sensitivity. If you have to rely on a human to do it, oftentimes they don't do it properly. Perhaps artificial intelligence in that fashion is good. And of course, the new forms of IMV, IMV(2), IMV(3), IMV(4), and IMV(5)[1,2] are important because they allow the ventilator to let the patient take control of the timing of breaths and also get the mandatory breaths out of the way unless they're needed. In addition, we have targeting schemes like servo targeting where the inspiratory effort makes the inspiratory pressure proportional to inspiratory effort so that you don't have as much work shifting.[3] Work shifting, as you're going to point out, can be a huge problem. And the only way to get around that is modes like NAVA or proportional assist or even something as simple as tube compensation that allow the patient to have more support when they demand it. Every other form of targeting causes the patient to work harder than necessary. But were those the issues that you wanted to cover?

Mireles-Cabodevila: Yeah, as we think about the trying to simplify the issue and thinking about just pressure support versus PAV in the context of over- or under-assistance, how would you put those to use? Katerina [Vaporidi], you have done a lot of work with PAV, so can you comment on adjusting the features that play into it as we encounter under-assistance or over-assistance?

Vaporidi: Proportional modes decrease your ability to passively ventilate a patient you have on an assisted mode. With pressure support and excessive assistance, the patient can just trigger the ventilator and then the ventilator delivers all the volume. This cannot happen with the proportional mode. However, you can provide more assistance than needed. You can have 80% assist when the patient would do some more effort with 50% assist and would be just fine. Therefore, you limit excessive assistance, but not all of it.

Chatburn: That's a really good observation. The ventilator mode taxonomy[4] defines a ventilator as a device that supports some portion of the work of breathing. But there's only one mode that we actually set the patient–ventilator work distribution, and that's PAV. It shows on the display how much work the patient does and how much the ventilator does. So that right there is the work shifting index, right?

Vaporidi: It's not very accurate though.

Chatburn: Yeah, I'm sure.

Vaporidi: Can I ask, you said that spontaneous breath is patient-triggered and cycled.

Chatburn: Yes.

Vaporidi: So what is the name for breath that is patient-triggered, but ventilator-cycled? The assist control breath?

Chatburn: Ventilator manufacturers and some authors call that an assisted breath. But that is an obsolete use of the word assist, which the ventilator mode taxonomy[4] defines as support of some portion of the work of breathing, and having nothing to do with triggering; certainly a ventilator is assisting the breathing of a paralyzed patient. But to answer your question, the mode taxonomy does not require a name for a breath that is patient-triggered and ventilator cycled, or for that matter one that is ventilator triggered and patient-cycled. The mode taxonomy is based on only two types of breaths, spontaneous (patient triggered AND patient cycled) and mandatory (machine-triggered OR machine cycled). Your questions implies that there are three kinds of mandatory breaths, machine triggered and machine cycled, machine triggered and patient cycled, and patient triggered but machine cycled. These may have some clinical significance and perhaps naming and displaying them on a ventilator could be useful. But that level of distinction is just a nuisance for classifying modes.

Mireles-Cabodevila: It's a mandatory breath. . .

Chatburn: Right. It's important to understand that if the ventilator gets involved in triggering or cycling, then the breath is classified as mandatory.

Vaporidi: Thank you for doing all this work in taxonomy.

Beitler: Thank you for that fantastic talk. You started your presentation with a modern history of some of the technology, and I can't help but feel that we're in an era of very rapid advancement of certain kinds of technology. Obviously, a lot of the focus now is on artificial intelligence (AI). So pertinent to your talk, do you foresee in the next 20–25 years that we're moving toward a post-taxonomy ventilator design where it's completely a black box like a lot of AI is? I would be skeptical of that except that ventilator companies are already starting to move in that direction. There are ventilators being developed and put into use now in certain regions where the clinician enters the diagnosis and perhaps a couple of parameters, and the ventilator does the rest: selects the mode and performs the titration over time. Do you see that being the future? That approach has clear advantages: It's potentially individualized ventilation in practice environments where there might not be the staffing available, or maybe the expertise if you will, but I think

there are a lot of risks with that too. And I was wondering if you could speak a bit about that.

Chatburn: I'm glad you asked that. I have thought about that a lot. And as you say, outside the United States, there are fewer restrictions on the use of AI in ventilators. If you think of where we're going in our daily life, AI's taking over everything. So I think certainly the trend is toward more automation in modes of ventilation. I wrote several years ago that I think eventually what we'll have is a ventilator that's a little box with only the pneumatic components. It's going to have a display that says, "connect the patient," "disconnect the patient." I think the resistance to AI in mechanical ventilation by the FDA is kind of hard to justify. You know, if your car can drive you to work without you touching the steering wheel, but you can't let the ventilator manage the patient, there's something wrong with that. You can only kill one person at a time with a ventilator. A car can be a weapon of mass destruction and yet we have fully self-driving taxis. Is there danger to that? Of course. But I think what's going to motivate more use of AI is what you mentioned, the lack of staffing, the lack of training at the bedside, the lack of ability to deal with the exponential growth in the number of tools in the toolbox.

Hess: I think about your tools in the toolbox, and my question is, who put the tools in the toolbox? Did manufacturers put the tools in the toolbox, or did clinicians and physiologists put the tools in the toolbox? Because it seems to me that many times new modes come about from engineers rather than from clinicians, and there are really only a handful of modes that are available on all ventilators. You mentioned dozens of modes that are out there. It seems to me that if all of those dozens of modes were so great and they showed improved patient outcomes they would be available on all the ventilators, not just one ventilator. If you want PAV, it's one ventilator. If you want NAVA, it's one ventilator, and you could go down the list. It really seems to be very technology-driven rather than clinically driven. Sometimes it is a solution in search of a problem.

Chatburn: I'm glad you brought that up because it takes a team to really make a mess, right? I have for many years been a consultant in industry helping manufacturers get their ventilators and their modes accepted through the FDA. I have often talked to the engineers that actually design ventilator. I'll say to them, why did you do this? Why do you have so many modes in your ventilator? And they said, well, because our clinical experts said they want that. And I will say, well, why did you do it this way? Like why did you make a ventilator mode where the patient can't possibly trigger a breath? Why would you do that? And they say, well, there's some doctor over in Europe that likes to put the ventilator on the patient when they're doing CPR and they don't want to trigger the breath with the chest rise effect. So the bottom line is it's driven—like everything else—by the money, how much you can sell. Now give me your best guess. What do you think the maximum number of modes is on a ventilator today? You'll be blown away. I counted 78 on one. And guess how many were mentioned in the manual? Sixteen. So you might argue that not only are there too many tools in the toolbox, many of them have no official name and no evidence of practical utility.

Hess: The technology exceeds our capability to use that technology safely at the bedside.

Chatburn: Which gets right back to what Jeremy [Beitler] said. If we're going to use this technology, if there's any value in it at all, then something better than the average person at the bedside has to be adjusting the settings and making the observations of patient outcomes. I suspect that most clinicians believe they don't have time, interest, or resources to understand the tools, much less document how effective they are. Seems to me that for us to make any real progress in this direction we must use AI, both for ventilator management and data mining of electronic medical records.

Beitler: I apologize, this diverges a little bit from the original talk, but I wonder if part of the problem is the way industry is set up. Ventilator modes are increasingly becoming proprietary, and manufacturers sometimes won't tell anybody what exactly the mode is doing, what the proprietary algorithm is that's being applied. I think that could lead to safety concerns, this divergence from transparency with clinicians. Maybe part of the solution is we need to be thinking of ventilators more like computers. There's hardware and there's software. You can license software on whatever device you have. I think that's the only way we'll really advance care at the pace that the field is advancing. When a hospital buys a ventilator it becomes a semi-permanent part of their fleet for about 20 years until they buy the next round because it's such a big investment. Even if these companies are making major advances, they are not going to have much meaningful impact on patient care until the way the industry is set up accommodates a completely different model, or for a generation of ventilator purchases to turn over. Could you share your thoughts about that issue?

Chatburn: I think the current paradigm of capitalism is to build ventilators with modes so that you can market all of them with different names—because that's product differential—rather than showing how things are the same. That has to change if we're ever going to move forward with this. And again, as you indicated, the taxonomy should become irrelevant at some point. I think ventilator

software should exist in the cloud as some form of AI that learns from its experience with all medical equipment netted together by the Network of Medical Things. The central server could learn from every patient who is ever ventilated. It seems like we're moving in that direction, that we have the technology to do that. Maybe not large language models alone but combined with reasoning models and similar AI technology. And I think when we get to that point, then a human clinician at the bedside fiddling with knobs on the ventilator and wondering what mode is in use will be ancient history. Now, if we don't progress, we'll just go off in the weeds and struggle for a long time with unused or misused technology, and God only knows what will happen.

Branson: Rob, as I usually tell you, it's not helpful to explain to physicians and respiratory therapy students how a ventilator works by comparing it to an electric circuit, because most don't know how an electric circuit works. Just like when you explain to somebody that PAV is like power steering, if they don't know how power steering works, then how can they understand PAV? But I think it goes back to my thoughts after Thomas's [Piraino] presentation, which is the role of the respiratory therapist at the bedside is interpreting this complicated technology and reducing it to practice. And I think that's important for our readers to understand. I agree, closed loop control ventilation and oxygenation is going to happen, but the intellectual properties surrounding it, the regulatory requirements which you know very well from FDA, nobody's going to share anything with anybody else. And Jeremy, I like what you said, and I think we should all adopt this idea. Ventilator manufacturers often come out with things that are new. I wouldn't call them an advancement necessarily. They're just new. And that goes to Dean's [Hess] point, which is that we often see 'new' features on a ventilator with no science whatsoever. And I think we have to distinguish between what is new and what is an advancement.

Chatburn: I appreciate your comments about the electrical circuit analysis. But our readers are not only physicians and therapists. You know, engineers understand all this. It's common sense to them. If we are going to use all the tools they have provided us then we need to understand a little about their worldview. And I hope my talk showed that you only need some very simple circuit analysis concepts to get a big return in comprehension. Perhaps by showing that we know their language, they will better know ours, and then make tools that are easier for all of us to understand.

Branson: Engineers do understand things, but during COVID, engineers created a ventilator because we needed rapid response ventilators. They just didn't include any capacity to deliver oxygen through it. Because it was a ventilator. Which just *ventilated* the patient. So, you're right it takes a team.

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